# Functional Anatomy of the Arm for Muscle Forces Estimation

Charles Pontonnier, Georges Dumont

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**C. Pontonnier[1], G. Dumont[2]**

## 1. ABSTRACT

This article presents how we use functional anatomy to develop a simplified model of the arm usable to estimate muscle forces. The model is developed as a motion analysis tool. The first application is the muscle forces estimation of the flexion/extension and pronation/supination of the forearm joints. This estimation is based on an inverse dynamics method, improved with additional constraints such as co-contraction factor between flexors and extensors of a joint. The article first presents the context of the study, and then presents the biomechanical model developed. Then we present the estimation of muscle forces step. Some results obtained for samples movements of the arm are presented and discussed. At last we conclude on the method and the perspectives that it presents.

## 2. INTRODUCTION AND CONTEXT

The proper design of workstations has a direct impact on the working conditions. Ergonomics takes an important place as a factor of productivity. Indeed, musculoskeletal troubles are often associated with the working stations in the industry, damaging the health of the worker and decreasing the productivity. In France, musculoskeletal troubles represent 75 percent of the professional diseases [1]. The purpose of this article is the presentation of a biomechanical model of the upper extremity that could be usable to decrease the musculoskeletal troubles. One of the solutions to improve the working conditions is to estimate and analyze muscle forces developed by a human during its work tasks. This estimation is a good way to improve the ergonomics [2], because of the tools that can be used to perform the analysis [3].
Biomechanical models of the arm are more and more complex, because of the addition of several degrees of freedom [4][5]. This complexity ensures a good behaviour of the model and a good accuracy in the reconstruction of the motion. Also, ergonomic postural evaluation techniques are based on functional anatomy [6][7]. As we want to add a dynamical criterion to the postural techniques, such as the Muscle Activity Envelope [8] or the Muscle Fatigue [2], in order to obtain information on the quality of the motion realized by a worker, we have to develop a model that uses functional anatomy. On the other way, in order to use these criteria during the conception of a working station, we have to develop a faster and simpler model of the arm, usable in real time or in a very short time.
The reminder of this article is structured as follows: first we present the biomechanical model developed in accordance with functional anatomy, then we explain the method

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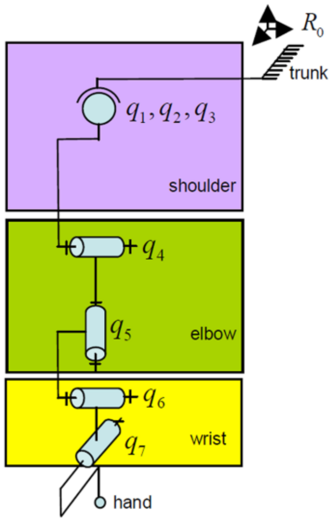
used to estimate muscle forces. Some results are presented and discussed for a sample motion. At last we conclude and present the perspectives of this work.

# 3. BIOMECHANICAL MODEL

## 3.1 Kinematical model

This kinematical model is based on functional anatomy [9]. Fig. 1 presents the model and the joint limits.




| *Joint* | *Function* | *Limits (°)* |
|---|---|---|
| $q_1$ | Flexion/Extension of the arm | **[0,180]** |
| $q_2$ | Abduction/Adduction of the arm | **[-60,90]** |
| $q_3$ | Internal rotation of the arm | **[-90,30]** |
| $q_4$ | Flexion/Extension of the forearm | **[0,150]** |
| $q_5$ | Internal rotation of the forearm | **[0,165]** |
| $q_6$ | Flexion/Extension of the hand | **[-80,90]** |
| $q_7$ | Abduction/Adduction of the hand | **[-50,20]** |

**Fig. 1: kinematical model and joint limits**

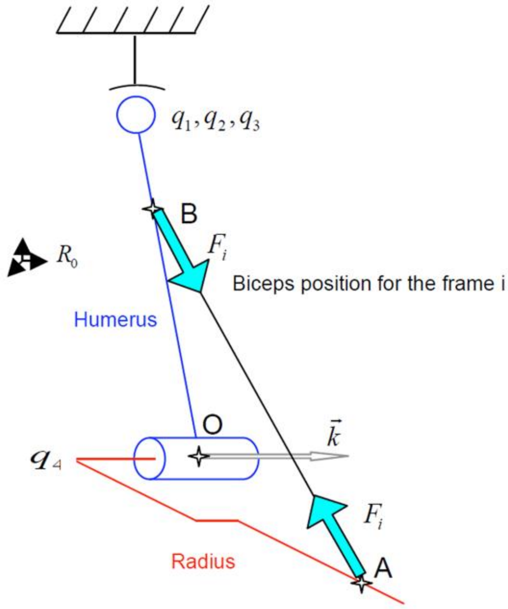
Our study is based on the elbow motion and the following joints in the kinematical chain. We consider the shoulder as a spherical joint. We do not take into account the constitution of the joint; we only use it to place the arm in the good position and orientation. For our range of motion, the spherical approximation is sufficient in order to rebuild the motion. In a further development we may use the Maurel's model [10]. The elbow is modelled as a gimbal joint that allows the flexion/extension and the internal rotation of the forearm. Elbow flexion/extension is considered as a single degree of freedom joint, allowing one rotation. Internal rotation of the forearm (pronation/supination) is represented as single degree of freedom as described in figure 4, following the mechanical axis defined by the head of the radius and the radio-ulnar joint. At last, the wrist is represented by two rotations that correspond to the flexion/extension and to the adduction/abduction of the hand.

## 3.2 Dynamical model

The dynamical model that we have developed is realized by using the Matlab© Simmechanics toolbox. The right arm is modelled as an articulated system of rigid bodies. The inertia parameters are defined for each motion capture subject with respect to the De Leva tables [11] and are automatically scaled with a pre-computing algorithm that we have developed [12]. The rigid bodies are connected with perfect mechanical joints with respect to the kinematical model (Fig. 1).

## 3.3 Musculoskeletal model

We have chosen to define the muscles as viscoelastic actuators. This approach is commonly accepted in the biomechanical community. We can cite for example the work of Zajac [13]. This model is useful to define the cost function to minimize and the constraints associated that allow us to estimate muscle forces, as it is described further. At each frame we rebuild the muscle origins and insertions in the main coordinate system to obtain the moment arms related to each joint.



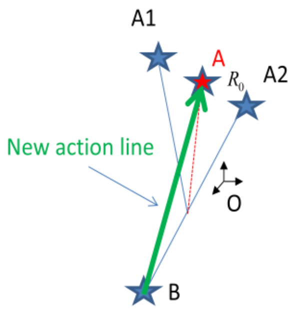

Contribution of the biceps for the elbow flexion/extension torque:

$$F_i R_i = [\overrightarrow{OA} \wedge F_i \frac{\overrightarrow{AB}}{\|\overrightarrow{AB}\|}].\vec{k}$$

With A origin of the biceps, B insertion of the biceps and O center of the elbow flexion/extension joint.

**Fig.2: Contribution of the biceps for the elbow flexion/extension torque**

Fig. 2 shows how the contribution of the biceps is computed for the flexion/extension of the elbow motion. The muscle is considered as a mechanical actuator and the force developed by the muscle is oriented in the main direction of the muscle. Furthermore, muscles work only in contraction. The origins and insertions of the muscles are defined on the basis of Clinically Oriented Anatomy [9]. The origins and insertions are scaled on the basis of the humerus and radius length. The computation of the muscle moment arms leads to the definition of equilibrium constraints between the computed torques issued from the inverse dynamics step and the muscle contribution for each joint, as it is described in the next section. Table 1 summarizes the adimensional coordinates of the muscles involved in the flexion/extension and the internal rotation of the forearm. Muscles with multiple origins and insertions are modelled with only one origin and one insertion per joint.




$$\overrightarrow{OA} = \frac{\overrightarrow{OA_1} F_{\max_1} + \overrightarrow{OA_2} F_{\max_2}}{\sum_m F_{\max_i}}$$

$$F_{\max} = \left( F_{\max 1} \frac{\overrightarrow{BA_1}}{\|\overrightarrow{BA_1}\|} + F_{\max 2} \frac{\overrightarrow{BA_2}}{\|\overrightarrow{BA_2}\|} \right) \frac{\overrightarrow{BA}}{\|\overrightarrow{BA}\|}$$

**Fig. 3: Definition of the action line and the maximum effort for a muscle**

Multiple origins are designed as a unique attachment defined as a barycentre. Maximum efforts are as well designed as the sum of the multiple maximum efforts in the direction of the new action line (Fig. 3).

| Muscle | Origin (adimensional) | Insertion (adimensional) |
|---|---|---|
| Biceps | $[-0.085\ -0.043\ 0.043]_{^iR_{shoulder}}$ | $[-0.045\ 0.045\ -0.099]_{^iR_{elbow}}$ |
| Triceps | $[0.043\ -0.021\ 0.128]_{^iR_{shoulder}}$ | $[0.062\ 0\ -0.037]_{^iR_{elbow}}$ |
| Brachialis | $[-0.037\ 0\ -0.531]_{^iR_{shoulder}}$ | $[-0.05\ -0.025\ -0.124]_{^iR_{elbow}}$ |
| Brachioradialis | $[-0.043\ 0\ -0.798]_{^iR_{shoulder}}$ | $[0\ -0.099\ -0.943]_{^iR_{elbow}}$ |
| Anconeus | $[0.043\ 0.085\ -0.957]_{^iR_{shoulder}}$ | $[0.037\ 0.025\ -0.124]_{^iR_{elbow}}$ |
| Pronator teres | $[0.043\ 0.106\ -0.957]_{^iR_{shoulder}}$ | $[0.05\ 0.025\ -0.298]_{^iR_{elbow}}$ |
| Pronator quadratus | $[0.037\ 0.05\ -0.868]_{^iR_{elbow}}$ | $[0.037\ -0.074\ -0.868]_{^iR_{elbow}}$ |
| Supinator | $[-0.198\ 0\ 0.05]_{^iR_{elbow}}$ | $[0\ 0.099\ -0.55]_{^iR_{elbow}}$ |

**Table 1: Adimensional origins and insertions of muscles**

## 4. MUSCLE FORCES ESTIMATION

This part presents the method we developed to obtain muscle forces estimation from motion capture data in three steps. For more details about the first two steps, please see [12][14]. Our method is an inverse dynamics method, using motion data (issued from motion capture) in order to estimate muscle forces.
The first step is an inverse kinematics step. To obtain joint positions, we resolve a reduced system of equations issued from the equality between the real orientation matrixes computed from markers positions and a combination of the rotation matrixes issued from the kinematical model. The main advantage of this method is its computation time, which is near 1 ms for one frame treatment.

$$
\begin{aligned}
&\text{For each frame :}\\
&\min f(F) = \sum_m \left( \frac{F_i}{(F_{\max})_i} \right)^2\\
&F = (F_1, F_2, F_3, F_4)\\
&\text{Under constraints :}\\
&h_1(F) = c - \sum_m F_i R_i = 0\\
&h_2(F) = F_1 + F_3 + F_4 - \alpha_{q_4}[F_2 + F_5] = 0\\
&h_3(F) = F_6 + F_7 + \alpha_{q_5} F_8 = 0\\
&g_i(F_i) = F_i - (F_{\max})_i \le 0
\end{aligned}
\tag{1}
$$

The second step is an inverse dynamics step. This model allows us to compute the joint torques involved in the motion. A numerical derivation is applied to the joint positions obtained from the inverse kinematics step. The simulation applies the Newton laws of dynamics to the system and thus computes the joint torques. These torques are then used to compute the muscle forces.

At last, we compute muscle forces using an optimization algorithm. The method consists in minimizing a cost function –defined as the global muscle fatigue. We use the joint torques to define equilibrium between muscle contributions for a joint and these torques. We developed some additional constraints in order to obtain physiologically realistic forces. The first ones are inequalities constraints that ensure that muscle forces do not reach the maximum capability of each muscle. We use the Zajac's model to define the maximum capability. The second one is an equality constraint that rely flexors activity of a joint to the extensors activity with a co-contraction factor, as defined in [14]. The optimization problem is resumed in equation 1.

## 5. RESULTS AND DISCUSSION

In this part are presented some results for a sample motion of the arm, a simple extension of the elbow. Results are presented Fig. 4.

The first four curves presents the joint positions and the related torques issued from inverse kinematics and inverse dynamics. Those results show that the joint torques are pretty weak during the motion. Actually, the subject does not hold any additional load (use a tool, hold a box). As we can see on the figure, the motion starts in a flexed position of the elbow ($q4$ =90°) and ends in a extended position ($q4$ = 5°). The forearm is in an intermediate position between pronation and supination. The whole motion is not presented here.

The last eight curves presents muscle forces estimated during the motion. The solution found by the optimization algorithm is not totally realistic. For example, triceps does not reach a peak of activity at the end of the motion, as it could be seen if we analyze EMG data of an extension of the elbow. We think that the co-contraction factor is not sufficiently accurate to perform realistic results. On the other way, the range of activity of any muscle is realistic and the global activity for each joint is representative and realistic. In other terms, some results such as these results are sufficient in order to use them in an ergonomics application.

The mean computation time is about 0.04 second per frame treatment. It is not sufficient to run in a real time application, but it is also faster than any other musculoskeletal simulation (OpenSim [15], AnyBody [16]…). We are working on a new estimation model in order to decrease the computation time. The objective is to reach 0, 01 second per frame (100 Hz).

At last, we want to improve the co-contraction factor by modelling it from EMG data and validate our results with EMG data in order to finalize the model.

## 6. CONCLUSION

In this article we have presented a biomechanical model of the arm usable for ergonomics applications. The model is divided in three parts corresponding to three analysis steps: the kinematical model, the dynamical model and the musculoskeletal model. Our muscle forces estimation is performed with an optimization algorithm, with additional constraints that ensure a sharing of the forces between flexors and extensors of a joint. Our results show that these constraints have to be improved and this is a field of research that we are currently exploring. We are doing some experiments including EMG measurements in order to improve the definition of the co-contraction factor and to validate at last the estimation.

Also, global results present a good behaviour for each joint and we want to use these results in ergonomics applications in our next works. At last, we are working on a new interpolation method that can meet the real time constraint.

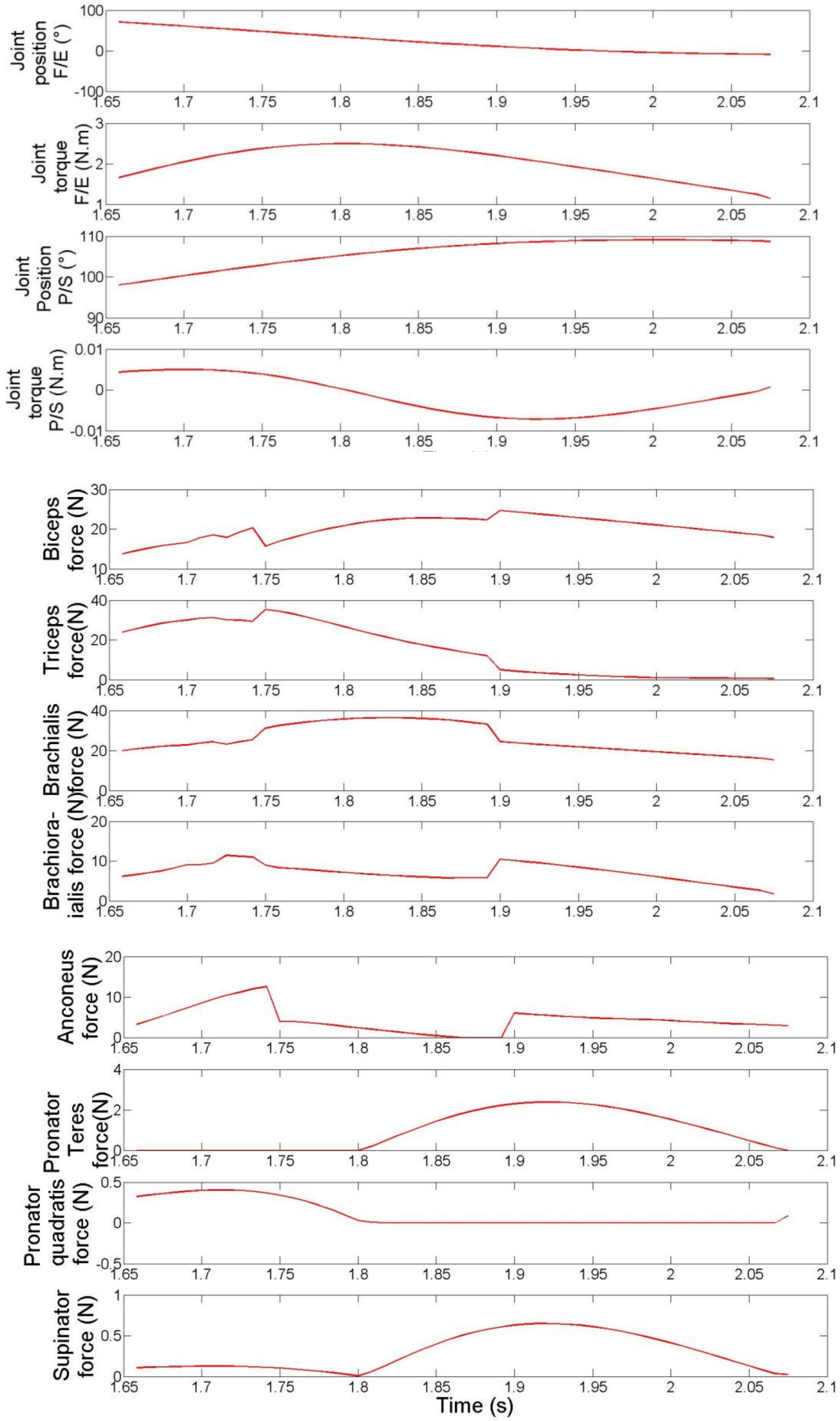



**Fig. 4: Results for a sample motion (extension of the elbow)**

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